(54) Title: NON-CONTACT SCANNING LASER SYSTEM

(57) Abstract

A refractive laser system is disclosed for use in refractive laser surgery and is a compact, low cost ophthalmic laser system (10, 35) which has computer controlled scanning for a non-contact delivery device for both photo-ablation and photo-coagulation in corneal reshaping. The advantages of the non-contact, scanning device (12, 37) used in the process include being safer, reduced cost, more compact and more precise. Lasers are selected with energies of 0.01-10 mJ, repetition rates of 1-10,000, pulse duration of 0.01 nanoseconds to a few hundreds of microseconds, and with spot sizes of 0.05-2 mm.

NON-CONTACT SCANNING LASER SYSTEM

1 BACKGROUND OF THE INVENTION

2 1. Field of the Invention

3   The present invention relates to laser ophthalmic

4 surgery using a compact, low-cost, low-power laser

5 system with a computer-controlled, non-contact process

6 and corneal topography to perform corneal reshaping

7 using either surface ablation or thermal coagulation.

8 This application is a continuation-in-part application

9 of Serial No. 07/985,617, filed December 3, 1992.

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12 2. Prior Art

13   Various lasers have been used for ophthalmic

14 applications including the treatments of glaucoma,

15 cataract and refractive surgery. For non-refractive

16 treatments (glaucoma and cataract), suitable laser

17 wavelengths are in the ranges of visible to near

18 infrared. They include : Nd:YAG (1064 nm), doubled-YAG

19 (532 nm), argon (488, 514 nm), krypton (568, 647 nm),

20 semiconductor lasers (630-690 nm and 780-860 nm) and

21 tunable dye lasers (577-630 nm). For refractive

22 surgeries (or corneal reshaping), ultraviolet (UV)

23 lasers (excimer at 193 nm and fifth-harmonic of Nd:YAG

24 at 213 nm) have been used for large area surface

25 corneal ablation in a process called photorefractive

26 keratectomy (PRK). Corneal reshaping may also be

27 performed by laser thermal coagulation currently

28 conducted with Ho:YAG lasers using a fiber-coupled,

29 contact-type process.   However, the existing

30 ophthalmic lasers as above described have one or more

31 of the following limitations and disadvantages: high

1 cost due to the high-power requirement in UV lasers
2 for photorefractive keratectomy; large size and
3 weight; high maintenance cost and gas cost (for
4 excimer laser), and high fiber-cost for contact-type
5 laser coagulation.
6   In light of the above, it is an object of the
7 present invention to provide ophthalmic laser systems
8 which offer the advantages of: low-cost, reduced size
9 and weight, reliability, easy-operation and reduced
10 maintenance. Another object of this invention is to
11 provide a computer-controlled scanning device which
12 enables use of a low-cost, low-energy laser for
13 photorefractive keratectomy currently performed only
14 by high-power UV lasers.
15   It is yet another object of the present invention
16 to provide a refractive laser system which is compact,
17 portable and insensitive to environmental conditions
18 (such as vibration and temperature). This portable
19 system may also be used for a mobile clinical center
20 where the laser is transported by a van. It is yet
21 another objective of the present invention to provide
22 a non-contact process for corneal reshaping using
23 laser thermal coagulation, where predetermined corneal
24 correction patterns are conducted for both spherical
25 and astigmatic changes of the corneal optical power.
26   The prior U.S. Patent No. 4,784,135 to Blum, et
27 al. and assigned to IBM teaches the first use of far
28 ultraviolet irradiation of a biological layer to cause
29 ablative photodecomposition. This patent teaches that
30 using a laser beam housing a wavelength of 193 nm and
31 an energy level of much greater than $10mJ/cm^2/pulse$ can
32 be used to photoablate corneal tissue without the
33 build up of excess heat. The present invention on the
34 other hand uses a process that allows the use of

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1   energy levels of less than 10 mJ/pulse in a process
2   that still allows photoablation.
3   There are several prior art U.S. Patents relating
4   to refractive surgery, or photorefractive keratectomy.
5   A UV solid-state fifth-harmonic of Nd:YAG (or Nd:YLF)
6   laser at 213 nm (or 210 nm), is disclosed in U.S.
7   Pat. No. 5,144,630 by the inventor, J.T. Lin. U.S.
8   Pat. No. 4,784,135 suggests the use of a UV laser with
9   wavelengths less than 200 nm, in particular Argon
10 Fluoride (ArF) laser at 193 nm, for non-thermal photo-
11 ablation process in organic tissue. Devices for beam
12 delivery and methods of corneal reshaping are
13 disclosed in U.S. Pat. No. 4,838,266 using energy
14 attenuator, and U.S. Pat. No. 5,019,074 using an
15 erodible mask. Techniques for corneal reshaping by
16 varying the size of the exposed region by iris or
17 rotating disk are discussed in Marshall et al,
18 "Photoablative Reprofiling of the Cornea Using an
19 Excimer Laser: Photorefractive Keratectomy" Vol. 1,
20 Lasers in Ophthalmology, pp. 21-48 (1986). Tangential
21 corneal surface ablation using ArF excimer laser or
22 harmonics of Nd:YAG laser (at 532 and 266 nm) is
23 disclosed in U.S. Pat. No. 5,102,409.
24   This prior art however requires high UV energy of
25 (100-300 mJ) per pulse from the laser cavity or
26 (30-40) mJ per pulse delivered onto the corneal
27 surface, where large area corneal ablation using a
28 beam spot size of about (4-6) mm which gives an energy
29 density of (120-200) mJ/cm^2. Moreover, the prior art
30 Argon Fluoride excimer lasers operate at a repetition
31 rate of (5-15) Hz and also limit the practical use of
32 the tangential ablation concept which takes at least
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1   (5-10) minutes for a -5 diopter corneal correction in
2   a 5-mm optical zone. The high energy requirement of
3   the currently used Argon Fluoride excimer laser
4   suffers the problems of: high-cost (in system,
5   erodible mask and gas cost), high-maintenance cost,
6   large size/weight and system are sensitive to
7   environmental conditions (such as temperature and
8   moisture).
9   The prior L'Esperance patent, US Pat. No.
10   4,665,913, disclosed the method of a scanning laser
11   for corneal reshaping. The proposed concept of this
12   prior art, however, had never been demonstrated to be
13   practical' or to achieve the desired clinical
14   requirement of smooth ablation of the corneal surface.
15   This prior art is not practically useful and had not
16   ever been demonstrated to be real because of the
17   conditions in the art. A high-power laser of (100-200
18   mJ) is required in the prior art in order to obtain a
19   useful beam with a substantially square spot size of
20   0.5x0.5 mm (see prior art, Col. 3, line 65 and Col. 4,
21   lines 1-14) due to the low efficiency of obtaining
22   such a beam, and which further requires a
23   substantially uniform density (see Col. 13, line 30
24   and Col. 15, line 25). To achieve myopic correction,
25   for example, the prior art (Col. 13, lines 61-66 and
26   Col. 15 lines 60-65) proposes a smooth laser density
27   increase with increasing scanning radius under the
28   condition that a substantially uniform density of the
29   scanning beam is required for a substantially uniform
30   scan area (Col. 15, lines 20-28 of L'Esperance).
31   Furthermore, L'Esperance teaches (Col. 4, lines 40-50)
32   that a depth of 0.35 mm in an area of 6 mm diameter
33   might be achieved in about 15 seconds when a beam spot
34   of 0.5x0.5 mm is used and each pulse ablated 14

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1   microns.   The prior art proposes the method of having
2   individual square beams (.05.x0.5 mm) scan to the
3   fashion of exact matching of the square boundaries to
4   cover the area of 6 mm, where the overlap among these
5   individual   beams   should   be   avoided,   otherwise
6   excessive ablation near the boundaries of each 0.5x0.5
7   mm spot causes ridges.   This is also part of the
8   reason that the prior art requires a substantially
9   square section of the individual beam with a
10   substantially uniform density.
11   The L'Esperance patent No. 4,665,913 requires a
12   complex apparatus to select a section of the beam
13   which is substantially uniform in density within a
14   substantially square spot "dot".   The overall
15   efficiency would be less than 10% from the output of
16   the laser window to the corneal surface and requires,
17   where a high power (at least 100 mJ) excimer laser
18   than will be required than the Blum, et al. patent.
19   It is almost impossible to match exactly the boundary
20   of each square beam to achieve a substantially uniform
21   scanned area even if each individual beam is perfectly
22   uniform and square in shape and the smooth increase of
23   the radius of scanned areas to obtain, for example, a
24   myopic correction profile, would still be almost
25   impossible to achieve for an overall smooth corneal
26   surface. The successive sweep of the scan areas would
27   always leave ridges between these sweeps.   It should
28   also be noticed that in L'Esperance's patent (Col. 18,
29   lines 10-28) uses overlaps between each of the scanned
30   areas to obtain the desired ablation profiles of
31   myopic (or other) corrections. However, the ridges
32   between each of the successive ablated areas are very
33   difficult to avoid if within each scanned area the
34   ablated profiles are not substantially uniform.   In

1 fact, one should expect a very rough surface on these
2 ablated areas in addition to the regular ridges
3 between each overlapped zones. One of the problems
4 found in these teachings is that each required
5 individual ablated area be substantially uniform and
6 in a round or square shape, which is very difficult to
7 achieve even if a perfectly uniform, square portion of
8 a fundamental beam is produced using a complex
9 apparatus for beam reshaping and having the high
10 initial power.
11 It is not clear that L'Esperance has found a
12 suitable scanning method or an effective method of
13 selecting a perfect beam (with uniform density and
14 well-defined shape) which would overcome the
15 above-described difficulties and make the proposed
16 teaching become practical in cost and design for any
17 clinical uses. In fact, L'Esperance's scanning method
18 has also been challenged by another prior art of
19 Muller, US Pat. No. 4,856,513, where the difficulties
20 and problems of L'Esperance's teachings are discussed
21 (see Col. 2, lines 1-40 of Muller's patent).
22 It is therefore a further object of the present
23 invention to provide a method and apparatus for
24 corneal reshaping by using software-driven new
25 scanning patterns which do not require substantially
26 uniform density or a specific spot shape. Contrary to
27 L'Esperance's teachings, which suggest that there
28 should be a perfect boundary match among each square
29 beams and that excessive overlap should be avoided,
30 the present invention proposes that a large portion
31 (50%-80%) of overlap among the individual beams is
32 necessary in order to achieve uniform ablated areas
33 and a smooth profile without ridges. Furthermore, a
34 low-power UV laser (0.1-2 mJ on corneal surface) at

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1   its bare-beam (having typically a 3-lop profile)
2   without any beam reshaping is sufficient to achieve a
3   smooth ablation surface based on the method proposed
4   in the present invention, where computer-controlled
5   beam overlap and orientation are employed.   In
6   addition to the surface quality problems, it is also
7   impossible for L'Esperance to achieve any meaningful
8   clinical results using his proposed techniques based
9   on the present low-energy laser of (2-4) mJ from the
10   output laser window and (0.1-2) mJ on corneal surface.
11   Therefore, another object of the present
12   invention is to provide a new method of beam scanning
13   which combines beam overlap and orientation for a
14   random beam density distribution on the ablated
15   corneal surface such that the individual beam profiles
16   are not critical, where the focused beam (spot size of
17   0.1- 1.2 mm) uses very low energy (0.1-2 mJ) and at
18   its bare-profile is delivered onto the corneal surface
19   in an averaged fashion.   Uniform, near flat-top
20   ablated areas of (1-9 mm in diameter) can be performed
21   by the nonuniform starting-beam, but only when a set
22   of specific predetermined overlap and orientation
23   parameters are used.   Portions of the theoretical
24   background was published by the inventor, J. T. Lin,
25   in SPIE Pro. vol 1644, Ophthalmic Technologies II
26   (1991), p.p. 266-275.
27   One of the essential feature of the present
28   invention for the photorefractive keratectomy process
29   is to use a scanning device in a laser system which
30   has high repetition rates, 50 to 50,000 Hz, but
31   requires less energy, ranging between 0.05-10 mJ per
32   pulse, or about 10 to 100 times less than that of the
33   prior art.   This new concept enables one to make the
34   refractive lasers at a lower cost, smaller size and

1   with less weight (by a factor of 5-10) than that of
2   prior art lasers. Furthermore, these compact lasers
3   of the present invention are portable and suitable for
4   mobile clinical uses. To achieve beam uniformity and
5   fast refractive surgery (30 to 60 seconds), a
6   mathematical model of the beam overlap and ablation
7   speed is also disclosed in the present invention.
8   For the laser thermo-keratoplasty (LTK) process,
9   the prior art uses fiber-coupled contact-type
10 procedure which involves the following drawbacks: (i)
11 slow processing speed (typically a few minutes to
12 perform eight-spot coagulation) which causes the
13 non-uniform collagen shrinkage zone; (ii) circular
14 coagulation zone which limits the procedure only for
15 spherical type correction such as hyperopia; and (iii)
16 the contact fiber-tip must be replaced in each
17 procedure.
18   In the present invention, a computer-controlled
19 scanning device is able to perform the laser
20 thermokeratoplasty procedure under a non-contact mode
21 and conduct the procedure many times faster than that
22 of the prior contact-procedure and without cost for a
23 fiber-tip replacement. Furthermore the coagulation
24 patterns can be computer predetermined for specific
25 applications in both spherical and astigmatic
26 corrections. The flexible scanning patterns will also
27 offer uniform and predictable collagen shrinkage.
28   For ophthalmic applications, it is another
29 objective of the present invention to include but not
30 limited to photorefractive keratectomy, laser
31 thermokeratoplasty, epikeratoplasty, intrastroma
32 photokeratectomy (IPK), phototherapeutic keratectomy
33 (PTK), and laser-assisted keratomileusis (LAK).
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1   SUMMARY OF THE INVENTION

2   The preferred embodiments of the basic ophthalmic

3   surgery method uses a laser system for the ophthalmic

4   surgery process, including: (1) a diode-pumped

5   solid-state lasers of Nd:YAG or Nd:YLF which is

6   frequency-converted by nonlinear crystals of KTP

7   (potassium titanyl phosphate), LBO (lithium

8   triborate), KNbO3 (potassium niobate) and BBO (beta

9   barium borate) into the fifth-harmonic at wavelength

10   of 213 nm or 210 nm with energy of 0.01 to 5.0 mJ; (2)

11   a compact, low-cost, low-power (energy of 1 to 10 mJ

12   per pulse) argon fluoride excimer laser at 193 nm; (3)

13   a frequency-converted Alexandite or Li:SAF or diode

14   lasers at (193-220) nm; (4) a compact, low-cost,

15   Q-switched Er:YAG laser at 2.94 microns; (5) a

16   free-running Ho:YAG (at 2.1 microns) or Er:glass (at

17   1.54 microns) or diode laser (1.9-2.5 microns); (6)

18   ultrashort pulse IR laser (750-1100 nm) and (7) mid-IR

19   (2.5-3.2 microns) laser generated from optical

20   parametric oscillation.

21   According to one aspect of the present invention,

22   the above-described basic lasers includes UV-lasers

23   (193-215 nm) and IR-laser (1.5-3.2 microns) which are

24   focused into a spot size of (0.05-2) mm in diameter,

25   where laser energy per pulse of (0.01-10) mJ is

26   sufficient to achieve the photo-ablation threshold

27   (PAT) energy density of 50 to 600 mJ/cm^2 depending upon

28   the laser parameters (wavelengths and pulse duration)

29   and tissue properties (absorption and scattering).

30   The prior art excimer laser uses large beam spot

31   ablation (4-6 mm) and require much higher laser energy

32   (100-300 mJ) than the low-power lasers presented in

33   this invention. In the present invention, a scanning,

34   non-contact device is used to control the low-power

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1   laser for corneal diopter change, whereas diaphragms
2   or masks are used in the high-power, high-cost excimer
3   lasers, and contact, fiber-tip is used in the
4   photo-coagulation procedure.
5   In another aspect of the present invention, a
6   mathematical model is presented according to the
7   optimal beam overlap for beam uniformity and fast
8   procedure and scanning patterns for refractive
9   corrections of myopia, hyperopia and astigmatism. For
10 high-repetition lasers (50 to 5,000 Hz as proposed
11 herein), refractive procedures may be completed in 20
12 to 60 seconds (depending on the diopter corrections)
13 in the present invention, where scanning speed is only
14 limited by the laser repetition rates.
15   A three-dimensional translation device (in X, Y
16 and Z) is integrated into the above laser systems,
17 where the laser heads are compact and light-weight and
18 can be steered to the corneal center by the
19 translation stages. The prior art high-powered excimer
20 laser systems are stationary and require a motorized
21 chair for corneal concentration. Beam steering and
22 scanning is very difficult for these high-power,
23 heavy-weight excimer lasers.
24   In yet another aspect of the present invention,
25 a free-running Ho:YAG (at 2.1 microns) or Er:glass
26 (at 1.54 microns) or diode (1.9-3.2 microns) laser
27 delivers a beam by a fiber waveguide and coupled to a
28 scanning device for non-contact procedure for laser
29 thermokeratoplasty (LTK), where optimal scanning
30 patterns for corneal coagulation are performed for
31 both spherical and astigmatic corrections.
32   In yet another aspect of the present invention,
33 the above-described laser system provides an
34 effective, low-cost tool for procedures of synthetic

1 epikeratoplasty (SEK), where the artificial lens is
2 sculpted with the laser to optimize lens curvature
3 without causing problems of corneal haze and
4 corrective regression. Real corneal tissues may also
5 be sculpted and implanted by the above-described laser
6 systems, a procedure known as laser myopic
7 keratomileusis (MKM). Furthermore the UV and IR lasers
8 disclosed in the present invention provide an
9 effective tool for phototherapeutic keratectomy (PTK)
10 which is currently conducted by high-power excimer
11 lasers and the procedure conducted by diamond-knife
12 called radial keratotomy (RK). This procedure
13 conducted by UV or IR lasers is called laser radial
14 keratotomy (LRK). The fundamental beam at 1064 or
15 1053nm wavelength of the present invention may also be
16 used for the intrastroma photorefractive keratectomy
17 (IPK), where the laser beam is focused into the
18 intrastroma area of the corneal and collagen tissue
19 are disrupted.
20   The ophthalmic applications of the laser systems
21 described in the present invention should include
22 photorefractive keratectomy, phototherapeutic
23 keratectomy, laser thermokeratoplasty, intrastroma
24 photokeratectomy, synthetic epikeratoplasty, and
25 laser radial keratotomy.
26
27 BRIEF DESCRIPTION OF THE DRAWINGS
28   Fig. 1 is a block diagram of computer-controlled
29 laser system consisting of a laser, scanning device,
30 power supply and the beam steering stage for
31 ophthalmic applications;
32   Fig. 2 is a block diagram for the generation of
33 ultraviolet wavelengths at 213 nm or 210 nm using
34 nonlinear crystals in a diode-pumped system;

1   Fig. 3 is a block diagram of a
2   computer-controlled refractive laser system of Ho:YAG
3   or Er:glass or diode laser in a non-contact scanning
4   mode for laser thermokeratoplasty;
5   Figs. 4A through 4E shows computer-controlled
6   scanning patterns for photo-coagulation in non-contact
7   LTK procedures for both spherical and astigmatic
8   corneal reshaping;
9   Figs. 5A and 5B are procedures for laser-assisted
10 myopic keratomileusis and hyperopic keratomileusis,
11 where the reshaping can be performed either on the
12 inner or outer part of the tissue;
13   Figs. 6A through 6D show computer-controlled beam
14 overlap and scanning patterns for myopic, hyperopic
15 and astigmatic correction using UV (193-240 nm) or IR
16 (0.7-3.2 microns) lasers;
17   Figs. 7A and B are laser radial keratectomy
18 patterns (LRK) using laser excisions for myopia
19 (radial-cut) and astigmatism (T-cut);
20   Figs. 8A through 8D show ablation patterns for
21 refractive correction using predetermined coatings on
22 UV or IR grade windows;
23   Figs. 9A through 9B show the spatial overlap for
24 uniform pattern;
25   Figs. 10A through 10B show the beam orientation
26 for smooth ablation; and
27   Fig. 11 shows the oriented expanding scanning to
28 achieve the required ablation profiles, where the
29 diameters are governed by a mathematical formula.
30
31 DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENT
32   The theoretical background of the present
33 invention with regards to the beam overlap and
34 ablation rate in photorefractive keratectomy,

1 intrastroma   photokeratectomy,   synthetic
2 epikeratoplasty,   phototherapeutic keratectomy and
3 myopic keratomileusis procedures described in the
4 present invention is as follows.

5   Given a laser energy per pulse of E (in mJ), an
6 intensity of I(in mJ/cm^2) may be achieved by focusing
7 the beam into an area of A, where I=E/A. For corneal
8 tissue ablation to occur requires the laser intensity
9 (I) to be above the photoablation threshold (PAT),
10 (60-120) mJ/cm^2 for UV-laser (193-215 nm) and (200-600)
11 mJ/cm^2 for IR-laser (2.5-3.2 microns). Therefore it is
12 always possible to tightly focus a laser beam and
13 achieve the PAT value even for a low-energy laser
14 (0.1-5) mJ. The drawback of using a low-energy,
15 small-spot laser for large area ablation is that the
16 operation time will be longer than that of a
17 large-spot but high-power laser. However, time of
18 operation   may   be   shortened   by   using   a
19 high-repetition-rate laser (higher than 50 Hz).
20 Small-spot, low-energy lasers for large area surface
21 ablation would becomes practical only when a scanning
22 device is used in a high-repetition-rate laser and
23 only when uniform beam profile can be assured by the
24 appropriate beam overlap. These two important issues
25 are addressed in the present invention.

26   The overall operation rate (R) for a given
27 diopter correction (D) is limited by the laser
28 scanning rate (R1) which is in turn limited by the
29 laser repetition rate. In addition, R is also
30 proportional to the tissue ablation rate (RT) which is
31 proportion to the laser intensity I (or energy
32 density) at a given energy E.

33   The diopter change (D) in the case of myopia is
34 related to the correction zone diameter (W) and the

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1   center ablation thickness (h0) and the ablation
2   profile h(x) (at corneal position x) by:
3        $h(x) = h0 + 1.32DX^2$ (1)
4        $h0 = -0.3315DW^2$ (2)
5   In a scanning system as disclosed in the present
6   invention, the number of ablation layers (M1) (without
7   beam overlap) required for D-diopter correction is
8   therefore related to the ablation thickness per pulse
9   (T1), D, and W by
10       $M1 = h0/T1 = -0.3315DW^2/T1$ (3)
11   To include the overlap factor (F), F=2 for a 50% beam
12   overlap scan and F=5 for 80% overlap, the required
13   effective number of overlapped ablation layers is
14   M1/F.
15   For a given ablation zone of W and laser focused
16   spot area of A, one requires an effective single-layer
17   scanning time (TS) of FW^2/A.
18   The total operation time (T) needed for h0 center
19   ablation or D-diopter correction becomes
20       $T = (M1/F)(TS)DW^4/E$ (4)
21       $T = DW^4/E$
22   Equation 4 gives us the scaling-law for operation
23   time required (T), the laser energy (E), diopter
24   change (D) and the ablation zone diameter (W). For a
25   given laser energy per pulse of E, the overall
26   operation rate (1/T) is independent to the laser
27   intensity (I) and beam spot size (A). By increasing
28   the laser average-power (P), defined by laser
29   energy/pulse X repetition rate, more total energy may
30   be delivered to the cornea per unit time. The
31   average-power (P) is the key factor which actually
32   determine the overall operation rate (or time)
33   required to achieve the diopter change. By realizing
34   that the scanning rate (1/TS) is proportional and

1   synchronized to the laser repetition rate (RP), we are

2   able to re-express Equation (4) as

3          $T = DW^4/P$ (5).

4   It is important to note that given an

5   average-power of P, the laser intensity must be above

6   the photo-ablation threshold(PAT) by either beam

7   focusing or increase the laser energy.

8   Based upon the above-described theory, some

9   important features are: (i) CW lasers (either UV or

10   IR) with low intensity normally can not cause

11   photo-ablation since the energy density is lower than

12   the PAT value; (ii) Lasers (UV or IR) at Q-switched or

13   mode-locked mode and with pulse-duration shorter than

14   100 nanosecond will normally achieve the intensity

15   above the PAT even at low-energy level of

16   0.05-5 mJ. In particular, picosecond lasers at high

17   repetition rate is desirable where energy in the

18   microjoule range would be sufficient. Moreover, the

19   Q-switched short pulse lasers have smaller thermal

20   damage than that of free-running lasers. The

21   cost-effective refractive lasers are those which have

22   high repetition rate (50 Hz and up) but operated at

23   low-energy (0.05-5 mJ) and short pulse duration

24   (0.001-20 nanoseconds). The preferred embodiments

25   disclosed in the present invention as discussed in

26   Fig. 1 are based upon this theory. Beam focusing and

27   scanning are always required to achieve the PAT and

28   smooth ablation profile. The individual beam profile

29   in the scanning system is not as critical as that in

30   prior art lasers which require a uniform overall

31   profile within the large ablation zone of (4-6) mm.

32   In laboratory tests, we have achieved a very smooth

33   ablation profile with zone diameter up to 8 mm

34   starting from a non-uniform focused beam profile which

1   was randomly scanned over the ablation zone of (1-8)

2   mm. Using overlap of (50-80)% of focused beam spot of

3   (0.2-1.5) mm, and a typical number of pulses delivered

4   to the corneal surface of 2,000-4,000, which assures

5   a sufficient beam overlap for smooth profile and

6   pulse to pulse energy fluctuation is not critical.

7   Referring to Fig. 1, a refractive laser system in

8   accordance with the present invention comprises a

9   basic laser 10 having UV (193-220 nm) or IR (0.7-3.2

10   microns) wavelength 11 coupled by a scanning device 12

11   having the beam from focusing optics 14 directed onto

12   a reflecting mirror 15 into target 16 which target may

13   be the cornea of an eye. An arming system 17 has a

14   visible wavelength (from a laser diode or He-Ne laser)

15   18 adjusted to be collinear with the ablation beam 11

16   and defines the centration of the beam onto the cornea

17   surface at normal incident. The basic laser head 20 is

18   steered by a motorized stage for X and Y horizontal

19   directions 21 and the vertical (height) direction 22

20   which assures the focusing beam spot size and the

21   centration of the beam onto the cornea. The system has

22   a computer controlled panel 23 and wheels 24 for

23   portable uses. The target 16 includes a human cornea

24   for applications of photorefractive keratectomy,

25   phototherapeutic keratectomy and laser radial

26   keratotomy (using the UV 193, 210, 213 nm or IR 2.9

27   microns beam focused on the corneal surface area) and

28   intrastroma photokeratectomy (using the 1064 or 1053

29   or 1047 nm beam, or their second-harmonic, focused

30   into the intrastroma area), and synthetic or real

31   corneal tissues for applications of synthetic

32   epikeratoplasty and myopic keratomileusis. The

33   computer controlling panel 23 also provides the

34   synchronization between the scanning gavo

17

1   (galvanometer scanner) and the laser repetition rate.
2   A commercially available galvanometer scanner made by
3   General Scanning, Inc. is used in scanning the laser
4   beam.

5   The laser systems described herein have been
6   demonstrated using photorefractive keratectomy
7   procedure with a diopter corrections up to -6 in PMMA
8   plasty and -12 in corneal tissues. In the case of
9   PMMA, we have also measured the diopters by a
10   lensmeter with well-defined readings in the ranges of
11   -1 to -12 diopters. This data provides the evidence of
12   predictable diopter corrections using the laser
13   systems of the present invention. Furthermore,
14   minimal tissue thermal damage of 0.3-1.0 microns were
15   measured by TEM (transmission electron microscopy). In
16   measurements, a multi-zone (MZ) approach for
17   high-diopter corrections (8-12) was used, where the
18   center zone is 3 mm and the correction power decreases
19   when the zone increases from 4 mm to 6 mm. This multi-
20   zone approach reduces the overall ablation thickness
21   and hence reduces the haze effect.

22   Still referring to Fig. 1, the basic laser 10,
23   according to the present invention, includes a
24   compact, optically-pumped (either flash-lamp or
25   laser-diode pumped) lasers of Nd:YAG, Nd:YLF or the
26   self-frequency-doubling crystal of NYAB (neodymium
27   yttrium aluminum) with pulse duration of 0.05-20
28   nanoseconds and repetition rate of 1-10,000 Hz. It is
29   known that this basic laser 10 is available using a
30   standard Q-switch or mode-lock, where the UV
31   wavelength at 209-213 nm may be achieved by the
32   frequency conversion techniques using nonlinear
33   crystals disclosed by the inventor in U.S. Pat. No.
34   5,144,630. The UV laser energy required for efficient

1   ablation ranges from 0.01 mJ to 5 mJ. The basic laser
2   also includes a compact, argon fluoride excimer laser
3   (at 193 nm) with repetition rate of (1-1,000) Hz,
4   energy per pulse of (0.5-10) mJ, pulse duration of
5   (1-50) nanoseconds and a compact, Er:YAG laser (at
6   2.94 microns) with repetition rate of (1-200) Hz,
7   energy per pulse of (50-500) mJ, pulse duration of
8   (50-400) nanoseconds and frequency-converted IR lasers
9   of diode laser, optically-pumped Alexandrite or Li:SAF
10 lasers, where efficient nonlinear crystals (as shown
11 in Fig. 2) may be used to convert the fundamental
12 wavelength (770-880 nm) into its fourth-harmonic at
13 the UV tunable wavelength of (193-220 nm) with energy
14 of (0.01-5.0) mJ, repetition rate of (1-10,000) and
15 pulse duration of (0.05-50) nanoseconds. Only two
16 nonlinear crystals are needed in this case and overall
17 efficiency is higher than that of the fifth harmonic
18 generation which requires three nonlinear crystals.
19 The basic laser may also include ultrashort pulsed
20 lasers, such as a commercialized mode-locked
21 Ti:sapphire laser or other solid-state laser, with
22 wavelength ranges of (750-1100 nm), repetition rates
23 of (0.01-100 MHz), energy per pulse of (0.01-100)
24 microjoules, and pulse durations of (0.05-10)
25 picoseconds where focused beam spot size of (0.05-0.5)
26 mm is required to achieve the ablation threshold.
27 When using an ultrashort pulse laser with very high
28 peak power density (gigawatts range), the tissue
29 ablation should be insensitive to laser wavelengths
30 since the tissue ablation is assisted by the plasma-
31 enhanced absorption with minimal tissue thermal
32 damage. A focused spot size of (0.05-0.5) mm of the
33 ultrashort pulsed lasers would be appropriate to
34 achieve the tissue ablation and precise ablation

1   profile is available by the scanning device proposed

2   by the present invention. Without a scanning device,

3   an ultrashort pulsed laser cannot be used in

4   refractive surgery due to its energy level of less

5   than 0.1 mJ and spot size smaller than 0.5 mm. The

6   above-described lasers may also be frequency-converted

7   into UV ranges of (190-220) nm suitable for

8   photoablation.

9   The basic laser also includes a mid-IR (2.5-3.2

10   microns) laser generated from optical parametric

11   oscillation (OPO) using a near-IR laser (such as

12   Nd:YAG or Nd:YLF, flash-lamp or diode-pumped) as the

13   pumping sources and KTP or BBO as the frequency

14   conversion crystals. The OPO laser has advantages

15   over the Q-switched Er:YAG laser, including higher

16   repetition rate (10-5,000 Hz) and shorter pulse width

17   (1-40 n.s.). These advantages provide faster surgical

18   procedure and reduced thermal damage on the ablated

19   corneal tissue. Typical energy per pulse of the OPO

20   laser is (0.1-10) mJ. Greater detail on OPO was

21   published by the inventor in Optical Communications,

22   vol. 75, p. 315 (1990).

23   Still referring to Fig. 1, the scanning device 12

24   is synchronized with the laser repetition rate, where

25   the computer software is capable of providing

26   predetermined patterns according to a patient's

27   corneal topography for the corrections of myopia,

28   hyperopia and astigmatism. Astigmatic correction, in

29   particular, is difficult to perform in prior art

30   systems using a non-scanning diaphragm but can be

31   easily achieved by the present invention using a

32   scanning device. Furthermore, a multi-zone procedure

33   for high diopter (6-15) changes can be performed by

34

1   the computer program rather than that of the
2   conventional mechanical iris.
3   The low-power laser systems described in the
4   present invention can perform the procedures normally
5   required in high-power lasers because a scanning
6   device is used to assure the uniform corneal ablation
7   by beam overlap and the ablation threshold is
8   achievable by small spot size.
9   Referring to Fig. 2, a preferred embodiment for
10 the basic laser 10 of Fig. 1 having a UV wavelength
11 includes a diode-pumped Nd:YAG (or Nd:YLF) 25 having
12 a fundamental wavelength of 1064 nm (or 1047 and 1053
13 nm) 26 and is focused by a lens 27 into a doubling
14 crystal 28 (KTP, KNbO3, LBO or BBO) to generate a
15 green wavelength 30 at 532 nm (or 524 and 527 nm).
16 The green beam 30 is further converted by a fourth
17 harmonic crystal 31 (BBO) to generate a UV wavelength
18 32 at 266 nm (or 262-263 nm) which is finally
19 converted by a fifth harmonic crystal 33 to generate
20 the UV wavelength 11 at 213 nm (or 209-211 nm). From
21 a commercially available diode-pumped Nd:YLF laser I
22 am able to achieve the UV (at 209-211 nm) energy of
23 0.01-2 mJ per pulse with average-power of 0.1 to 0.5
24 W. This energy level when focused into a spot size of
25 (0.1-0.5) mm is sufficient to ablate the corneal
26 tissue. This diode-pumped fifth-harmonic system
27 provides the most compact refractive UV solid-state
28 laser available today with the advantages of long
29 lifetime, low maintenance, portability and absence of
30 toxic gas in comparison with the excimer lasers
31 currently used by other companies. Furthermore by
32 using the fundamental wavelength at 1064 nm (or 1053
33 or 1047 nm) or their second-harmonic (at 532, 524, or
34 527 nm), intrastroma photokeratectomy procedure may be

1   performed by focusing the beam into the intrastroma
2   area of the cornea.   The laser presented in the
3   present invention provide a compact, portable and
4   low-cost IPK laser and has an advantage over the
5   lasers used by other companies where the systems are
6   currently more than five times heavier and are more
7   costly.
8   In Fig. 3, a commercially available Ho:YAG (or
9   Er:glass) or diode laser 35 (either flash-lamp or
10   laser-diode pumped) is coupled by a fiber optic
11   waveguide 36 with core diameter of (100-600) microns
12   to a scanning device 37, in which the fundamental beam
13   38 with a wavelength of 2.1 (or 1.54) or (1.9-2.5)
14   microns which is collimated by a lens 40 and coupled
15   to the scanning gavo 41 and focused by another lens 42
16   onto the beam splitters 43 and 44, and finally
17   delivered to a target (such as a patient's cornea) 45.
18   The IR (2.1 microns) laser beam 38 is collinear with
19   the aiming beam 46 (visible He-Ne or diode laser) and
20   the patent corneal center is also defined by a
21   commercial slit-lamp microscope station 47. The
22   above-described apparatus offers the unique feature of
23   non-contact laser thermokeratoplasty for precise
24   coagulation in both spherical and astigmatic corneal
25   power corrections with scanning patterns predetermined
26   by a computer software hereinafter discussed. The
27   focusing lens 28 may be motorized for varying the
28   focal point and thus varying the coagulation cone size
29   for optimal results. In the prior art of fiber-tip
30   contact system, the precision of the coagulation zone
31   and patterns are limited by doctors manual operation
32   which is a much slower procedure than the computer
33   controlled scanning device described in the present
34   invention. The requirement of replacing the fiber-tip

1   after each operation is also a drawback of the prior

2   art systems. The advantages of the present system

3   includes: precision coagulation zone and spot size,

4   flexible patterns for a variety of corrections, fast

5   processing time and elimination of the need for

6   fiber-tip replacement.

7   Still referring to FIG. 3, the basic laser 22 in

8   accordance with the preferred embodiment of the

9   present invention is a free-running or continuous-wave

10   (CW) flash-lamp or diode-laser pumped Ho:YAG (at 2.1

11   microns) or Er:glass (at 1.54 microns), or IR diode

12   laser (1.9-2.5 microns) with average power of 0.5-5 W,

13   pulse duration of 200-2,000 micro-seconds (if

14   free-running). In the present invention, the IR

15   wavelengths of 1.54 and 2.1 and (1.9-2.5) microns are

16   chosen due to their strong tissue absorption which is

17   required in the photo-coagulation processes. Similar

18   lasing media of Ho:Tm:YAG and Ho:Tm:Cr:YAG is also

19   included in the preferred embodiments of the present

20   invention. The CW diode laser (1.9-2.5 microns) may

21   be scanned in a faster rate than that of the free-

22   running lasers.

23   Figs. 4A through 4E summarize the possible

24   coagulation patterns suitable for both spherical and

25   astigmatic corneal reshaping in the LTK procedures in

26   a cornea 50. Fig. 4-A with coagulation zone (CZ) of 5

27   to 9 mm and spot number (SN) of (8-16) provides

28   hyperopic corrections of 1-6 diopters; Fig. 4-B has a

29   coagulation zone of 1-3 mm suitable for myopic

30   corrections; Fig. 4-C has radial coagulation zone and

31   spot number of 16-32, suitable for spherical hyperopic

32   correction; Fig. 4-D has a coagulation zone of 1-9 mm

33   and spot number of 50-200, suitable for precise

34   coagulation control to stabilize and reinforce the

1   collagen shrinkage tension; Fig. 4-E is designed for

2   astigmatic change, where the coagulation patterns are

3   chosen according to the corneal topography. By using

4   the computer-controlled scanning, these patterns may

5   be easily generated and predetermined according to the

6   measured corneal topography of each patients. A

7   combination of these patterns illustrated in Figs. 4-A

8   to 4-E enables the treatment of patent's optical power

9   correction in all aspects of myopia, hyperopia,

10   astigmatism and their mixed vision disorder.

11   Furthermore, laser parameters such as energy per

12   pulse, spot size and scanning patterns also provide

13   another degree of freedom for the laser

14   thermokeratoplasty process which are not usually

15   available in the prior art systems using the contact

16   fiber-tip.

17   The appropriate parameters relating to Fig. 4A-B

18   are: laser energy per pulse of 5-50 mJ for

19   free-running mode (200-400 micro-second duration),

20   beam spot size of (0.1-1) mm, laser repetition rate of

21   5-30 Hz, coagulation zone of (1-10)mm, spot number of

22   8-200 spots and fiber core diameter of 100-600

23   microns, for a flash-lamp-pumped system. Also

24   disclosed is the use of a diode-pumped Ho:YAG, either

25   in a pulse-mode or continuous-wave (CW) mode. For a

26   CW mode laser, energy of 10-100 mW is sufficient for

27   coagulation when spot size of 0.05-0.5 mm is employed.

28   In the diode-pumped system in CW mode or with a

29   high-repetition-rate 20-100 Hz, a fast scanning

30   enables completion of the coagulation procedures

31   within 2-20 seconds depending upon the coagulation

32   zone and spot number required. Fast scanning also

33   provides a uniform collagen shrinkage unlike that of

34   the prior art system using a manually operated

1   fiber-tip which normally takes 1 to 5 minutes to

2   complete in a multiple coagulation zone and high spot

3   number. It is difficult to use a manually operated

4   fiber-tip to generate the precise patterns as

5   illustrated in Fig. 4 which can be easily performed in

6   the computer-controlled scanning device as disclosed

7   in the present invention. The patient's eye motion and

8   decentration is a problem for prior art systems, but

9   it is not a critical factor in the fast scanning

10 device described herein.

11   Referring to Fig. 5, a laser-assisted myopic

12 keratomileusis (MKM) and hyperopic keratomileusis

13 (HKM) can be performed either on the outer corneal

14 surface 51 or on the inner surface 52 to reshape the

15 resealed corneal tissue without materially effecting

16 the Bowman's layer. The preferred lasers are

17 described in Fig. 1 including the UV (193-220 nm) and

18 IR (2.5-3.2 microns) lasers. The non-invasive

19 laser-assisted procedure disclosed in the present

20 invention has the advantages over the procedures of

21 photorefractive keratectomy and laser

22 thermokeratoplasty including being safer, more stable

23 with a higher diopter change, and without materially

24 affecting epithelium and Bowman's layer. In

25 comparison with the conventional keratomileusis, the

26 laser-assisted myopic keratomileusis and hyperopic

27 keratomileusis do not require corneal freezing and can

28 perform very high diopter change not available by

29 radial keratotomy or photorefractive keratectomy.

30 Laser-assisted corneal preshaping can also be employed

31 for a donor cornea in the procedure currently

32 performed by epikeratophakia. Details of conventional

33 lamellar refractive surgery may be found in Leo D.

34

25

1 Bores, Refractive Eye Surgery (Blackwell Scientific
2 Pub., 1993), Chapter 10.
3   Figs. 6A through 6D shows a nearly flat-top beam
4 profile achieved by overlapping a series of laser
5 beams, where the degree of overlap, 50%-80%, depends
6 on the individual beam profiles which are not required
7 to be flat-top.   In the present invention, the
8 preferred individual beam profile is either a 70%
9 Gaussian or a symmetric profile.   In the laboratory,
10 I have demonstrated a smooth laser-ablated PMMA
11 surface with zone diameter of 3-6 mm by overlapping a
12 large number of pulses, 500 to 5,000, each one having
13 a spot size of 0.8-1.2 mm. Moreover smooth transition
14 among the ablation zones were achieved without the
15 transition zone steps found in prior art systems using
16 mechanical diaphragms.   In addition to the myopic and
17 hyperopic scanning patterns of 6B and 6C, one of the
18 significant features of the present scanning device is
19 that it can generate predetermined patterns based upon
20 the corneal topography for astigmatism correction (see
21 6D).   Corneal scar may also be easily located by a
22 topography and photoablated by a laser based on the
23 computer-controlled scanning patterns. The preferred
24 lasers for the procedures described in Fig. 6 are
25 discussed in connection with Fig. 1.
26   Still referring to Fig. 6, the scanning schemes
27 were tested by ablation on PMMA plasty. The computer
28 software is based upon the mathematical model
29 described earlier in equations 1 and 2 where the
30 center ablation thickness was equally spaced to define
31 the associate scanning diameters. Given the ablation
32 thickness per pulse and per ablation layer (at a given
33 scanning diameter), one may easily obtain the overall
34 corneal surface ablation profile, (see equation (1)).

26

1   The number of required ablation layers is therefore
2   proportional to the diopter change (D) and square of
3   the ablation zone (W).   The computer parameters
4   designed in the present invention include: diopter
5   change (D), optical zone diameter (W), and the degrees
6   of overlap in both tangential (TD) and radial (RD)
7   direction of the scan patterns as shown in Figs. 6A
8   through 6D. Smooth PMMA surface ablation was achieved
9   by optimization of laser spot size, energy and the
10   overlap parameters of TD and RD.   Experimental data
11   indicates   that   larger   overlap   provides   smoother
12   surface ablation, however, longer ablation time is
13   required for a given diopter change, laser energy and
14   repetition rate (RR).   Larger RR, 50-100 Hz, provides
15   shorter ablation time which is typically in the range
16   of (20-40) seconds for diopter changes of 2-8 in
17   myopic treatment based upon my measurements.   The
18   prior art high-power excimer lasers with a typical RR
19   of 5-15 Hz will be impossible to achieve the results
20   described above even if they use the present scanning
21   device.
22   Still referring to Figs. 6, using the UV lasers
23   (193, 210 and 213 nm) I have achieved ablation depths
24   of (20-40) microns by overlapping (2000-4000) laser
25   pulses, which give an ablation depth of 0.05-0.1
26   microns per pulse. The ablation depths are measured
27   by 1a microsensor (made by Tencor Instruments) which
28   has a resolution of about 0.5 microns or better.
29   Ablation   curves,   ablation   depth   versus   laser
30   intensity, were obtained by varying the laser energy
31   or the spot size. Given the ablation rate (ablation
32   thickness per pulse), I am able to calibrate the
33   number of pulses and the degree of beam overlap
34   required to achieve the diopter change on the PMMA,

1 where the diopters of the ablated PMMA are measured by

2 the standard lensmeter.   In vitro measurement of

3 corneal tissue ablation can be calibrated according to

4 the comparison of the ablation rate between PMMA and

5 tissue. For myopic and hyperopic corrections, I have

6 used circular scanning patterns with beam overlap

7 controlled by the tangential scanning speed and

8 diameters of the adjoined circles.   The preferred

9 scanning scheme is from small circle to large circle.

10 For example, given a laser spot size of 1 mm, a radial

11 overlap of 50% will require the scanning circle to

12 start from 1 mm diameter to 5 mm diameters with an

13 increment of 0.5 mm for an optical zone of 5 mm.

14 Furthermore, a tangential overlap of 50% requires the

15 scanner to move at an angular speed of about 23

16 degrees within the interval between each laser pulse.

17 In my computer-controlled scanning device, software

18 was developed to synchronize the laser repetition rate

19 with the scanning gavo to control the above-described

20 overlap patterns.   In addition to the circular

21 patterns described for myopic and hyperopic

22 treatments, a linear scanning pattern may also be used

23 in particular for the myoptic and astigmatic

24 corrections.

25   It is important to note that a uniform individual

26 beam profile and energy stability of the laser, under

27 the present scanning device, are not critical in

28 achieving an overall uniform ablation zone whereas

29 they are very critical for prior art systems using

30 expanding iris devices. Given the ablation rate per

31 overlapped circle, the overall diopter correction may

32 be achieved by the appropriate increment in diameters

33 of the expanding circles. Greater details of beam

34

1   scanning and overlapping will be further discussed in
2   connection with Figs. 9-11.
3   Referring to Figs. 7A and 7B, a laser radial
4   keratectomy (LRK) performed by laser excision has
5   advantages over the conventional diamond-knife radial
6   keratotomy (RK) including higher predictability and
7   reproducibility by precise control of the excision (or
8   ablation) depth.   Furthermore, using the scanning
9   device of the present invention, laser radial
10   keratotomy may be performed easily and rapidly with
11   less dependance upon the surgeon's skill and
12   experience.   Corneal reshaping may be performed by
13   controlling the laser parameters such as spot size,
14   intensity, scanning speed, beam overlap, and the
15   excision depth per pulse which typically ranges from
16   0.2 to 0.5 microns.   The excision depth precision of
17   a laser is at least 10 times better than that of a
18   knife.   This "laser-knife" should be able to perform
19   all the radial keratotomy procedures performed by a
20   "diamond-knife" by using similar techniques to those
21   introduced in the Book of Leo D. Bores, Refractive Eye
22   Surgery, Chapters 8 and 9.   Examples of laser radial
23   keratotomy are shown in 7A for myopia (radial-cut) and
24   7B for astigmatism (T-cut).   The preferred lasers for
25   laser radial keratotomy include the lasers described
26   in Fig. 1.
27   Referring to Figs. 8A and 8D, the ablation
28   patterns suitable for refractive procedures may be
29   generated by using coated windows such as UV (or IR)
30   grade fused silica, MgF, BaF or sapphire (when an IR
31   laser is used), with preferred thickness of (0.5-2) mm
32   and diameter of (8-15) mm.   Referring to Fig. 8A,
33   scanning laser beams 53 (at wavelength of UV or IR)
34   with circular scanning pattern to deliver uniform (or

1 constant) laser energy over the coated window 44 with

2 coating specification (at UV or IR wavelength)

3 according to the profile on the corneal tissue 55 (or

4 PMMA surface) will also achieve the same pattern

5 described by equation (1). Figs. 8B and 8C show the

6 reflection profiles of the coated windows for myopia,

7 hyperopia and astigmatism, respectively, based on

8 predetermined diopter changes. These coated windows

9 disclosed in the present invention can be reused for

10 cost effectiveness and has an advantage over the prior

11 art system using the disposable mask which is costly

12 and is difficult to provide reproducible results due

13 to the non-uniform transmission or ablation properties

14 of the mask.

15   Greater detail of the features of the present

16 invention regarding beam overlap, scanning and

17 orientation in order to achieve uniform ablation

18 profiles to meet the clinical requirements of corneal

19 reshaping are demonstrated as follows. The actually

20 measured PMMA profiles were generated from the

21 Microsensor (made by TENCOR INSTRUMENTS, INC.) using

22 our ArF laser (the Compak-200 Mini-Excimer system,

23 made by LaserSight, Inc.) having laser parameters of:

24 (2-4 mJ) energy at the output window, operated at

25 (50-200) Hz, with the beam focused onto the corneal

26 surface at a spot size of about (0.2-1.2) mm, with

27 energy per pulse of (0.5-1.5) mJ, tunable by a coated

28 MgF window.

29   Referring to Fig. 9A, we show the schematic of

30 the motion of the scanning beam with a spot size of 1

31 mm in this example. Beam overlap function(L) is

32 defined by the beam displacement parameters of dx and

33 dy (in x and y direction, respectively, on the corneal

34 plane) adjustable by the computer controlled software,

30

1   where Lx=1-dx/R and Ly=1-dy/R, where R is the beam
2   diameter. The degrees of smoothness (DS) of the
3   ablated PMMA surface (a plastic sheet which has been
4   commonly used for the calibration of UV laser ablation
5   on corneal tissue) is governed by the degrees of
6   overlap function L=Lx + Ly. Greater DS can be
7   performed by using greater L, which, however, will
8   also cause a slower procedure speed (v), at a given
9   laser average-power(p), beam spot size(R) and
10 energy per pulse (E). Desired procedure time of 20 to
11 50 seconds are typical for patient diopter corrections
12 (myopic) of D=-3 to -10, where patient centration is
13 conducted by a visible fixation light for the patient
14 to look at without eye movement. Including some of
15 the compensation from the recovered epithelium filling
16 on the ablated corneal surface, the roughness of the
17 corneal tissue, calibrated by the PMMA surface, should
18 be within the range of (0.2-2) microns. Therefore, we
19 are optimizing the parameters of dx, dy, L,p, E and R
20 in order to achieve the above-described clinical
21 requirements.
22   Referring to Fig. 9B, a comparison is shown to
23 demonstrate the degrees of smoothness of the ablated
24 PMMA at two sets of displacements: curve A (dx=dy=0.5
25 m) and curve B(dx=0.5 mm, dy=0.3 mm). These PMMA
26 profiles were generated from a Microsensor scanned
27 along the y direction to show the difference in
28 smoothness caused by the difference in dy values (at
29 a fixed dx value). It is clearly demonstrated by
30 comparing Curves A and B that a smoother surface is
31 generated with a smaller displacement (dy=0.3 mm), or
32 larger beam overlap Lx=70%. In this particular
33 example, the basic beam profile is worse than a 50%
34 Gaussian and actually has a three-lop structure which

1   is typical in an ArF excimer laser. Even under this

2   poor beam uniformity condition, we are still able to

3   obtain very uniform overall ablated areas of (2-9) mm

4   in diameter, as shown in Fig. 9B (curve B) with

5   surface roughness less than 1 microns (vs. about 10

6   microns in curve A), when a set of appropriate beam

7   overlap parameters are used. Smaller dx and dy will

8   further improve smoothness, which, however, may take

9   a longer operation time. As shown in above example

10   (using dx=0.5 mm and dy=0.3 mm), only 30 seconds is

11   needed for a D=-4 diopter correction with enough

12   smoothness of the PMMA surface, where I used a pulse

13   energy of 0.9 mJ (on the PMMA surface), with the

14   system operated at 100 Hz in this example.

15   In addition to the overlap function, I have been

16   able to further improve the beam uniformity by the

17   beam orientation method as follows. As shown in Fig.

18   10A, I used linear scan patterns for multi-layer

19   ablation on a PMMA sheet, where parameters of E=0.9

20   mJ, spot size of 1 mm, dx=dy=0.5 mm were used. In one

21   case, I repeated the linear scan pattern along the

22   x-direction, or rotation angle (A)= zero, for about 25

23   times (layers). To see the improvement due to pattern

24   orientation, I tried the second case by rotating the

25   linear-scan angle (A) by about 65 degrees in each

26   successive scan layers. An angle A=65 degrees was

27   chosen in this particular example to randomize the

28   basic beam structure (having a non-uniform profile)

29   and to achieve the uniform overall ablation. This

30   averaging procedure by beam orientation will largely

31   reduce the potential roughness caused by the basic

32   beam structure, noting that rotation angles, such as

33   20, 30, 60 or 120 degrees (in which 360 degrees can be

34   divided into integers), should be avoided to prevent

32

1   repeated patterns after a few rotation layers. A
2   larger angle(A) is chosen for smaller diopter
3   corrections and vice versa for the best results. This
4   is to make sure that enough beam randomization is
5   performed for various diopter corrections which are
6   proportional to the numbers of scanned layers.
7   Comparisons are shown in Fig. 10B for A=0 (nonrotated
8   case, curve A) and for A=65 (rotated case, curve B),
9   where dx=dy=0.5 mm were used in both cases.
10   Significant smoothness of ablated PMMA was achieved in
11   the rotated case (curve B) even when a large
12   displacement of dy=0.5 mm was used, compared to curve
13   B in Fig. 10B and curve A in Fig.9B. The larger
14   displacement, or smaller overlap results in a faster
15   procedure, however, this results in a loss of
16   smoothness if beam rotation is not used. Using the
17   above-described techniques, I am able to generate the
18   predetermined ablation profiles corresponding to
19   various refractive corrections such as myopic,
20   hyperopic and astigmatic with clinically acceptable
21   tissue smoothness and procedures times requirement.
22   Referring to Fig. 11, an example for myopic
23   correction is shown. Fig. 11A shows the schematic of
24   rotated ablated areas with increasing diameters (from
25   about 0.5 to 6 mm) governed by Equation (1), where a
26   typical number of layers (or scanned areas at various
27   diameters) of 25 is needed for a -5 diopter
28   correction. For an optical zone of 5 mm, this
29   represents an ablation rate of about 2 microns in
30   corneal tissue in each layer, where a pulse energy of
31   about 0.9 mJ at spot size of 1 mm and repetition rate
32   of 100 Hz is used. For smaller diopter corrections,
33   a smaller energy (0.6-0.8 mJ), or smaller ablation
34   rate (0.5-1.0 microns) is desired for smoother and

1   more accurate results. Moreover, a smaller spot size
2   of (0.1-0.5 mm) may be used for better control of the
3   ablation profile (with greater accuracy), but a faster
4   laser repetition rate larger than 500 Hz would be
5   required for a reasonable procedure speed of (20-50)
6   seconds to cover (-3 to -10) diopter corrections. In
7   this situation the diode pumped UV solid state laser
8   described earlier will be a better candidate than the
9   Excimer laser.   Fig. 11B shows the PMMA ablation
10   profile measured from a Microsensor using the
11   techniques shown in Fig. 11A, where an ablation zone
12   size of about 5 mm with center depth of about 16
13   microns were shown.   I believe that the PMMA profiles
14   shown in Figs. 9 through 11 represent, for the first
15   time, the novel features of the techniques disclosed
16   in the present invention. Some of the prior art has
17   never demonstrated the actual ablation data, although
18   a simple concept of beam scanning has been proposed.
19   The comparisons in Figures 9 and 10 have demonstrated
20   that the prior techniques as set forth in the
21   background hereto would never achieve the smooth
22   surface as shown here. In addition, given the laser
23   parameters proposed in the present invention of
24   low-energy (2-4 mJ) with nonuniform basic beam profile
25   and without using mechanical beam re-shaping, it is
26   impossible for the prior art to achieve clinically
27   meaningful results. A high-power laser of 100-300 mJ
28   with a complex means of beam uniformity is always
29   required in the prior art patents.
30   The method disclosed in the present invention
31   combines beam scanning, overlapping and pattern
32   rotation (randomization) provides a powerful yet
33   simple technique for optimal results of laser
34   refractive surgery which involves both clinical

34

1 aspects (ablation diopter, ablation optical zone,

2 smoothness, patient centration and operation speed)

3 and engineering aspects (beam profile, uniformity,

4 stability, energy, spot size and delivery systems).

5   It is worth emphasizing that the concept of

6 achieving a smooth ablation surface by using the

7 randomly rotated scanning pattern as disclosed in the

8 present invention would not be demonstrated if the

9 microsensor were not used to measure the PMMA

10 profiles. I have preformed hundreds of PMMA profile

11 analyses at various laser parameters together with the

12 theoretical model presented in equations (1) - (5) are

13 the key factors behind the present process.

14 Furthermore, the refractive correction profile,

15 governed by equation (1) would be very difficult to

16 justify after the scanning method is applied to the

17 target (PMMA and corneal tissue) if the microsensor is

18 not available to the user. The PMMA data presented in

19 the present invention have also been employed on

20 corneas, where hundreds of patient's have been treated

21 by the Compak-200, Mini-Excimer with predictable power

22 corrections and smooth tissue ablation. Clinical

23 results are to be presented in ophthalmology

24 conferences.

25   While the invention has been shown and described

26 with reference to the preferred embodiments thereof,

27 it will be understood by those skilled in the art that

28 the foregoing and other changes and variations in form

29 and detail may be made therein without departing from

30 the spirit, scope and teaching to the invention.

31 Accordingly, the method and apparatus, the ophthalmic

32 applications herein disclosed are to be considered

33 merely as illustrative and the invention is to be

34 limited only as set forth in the claims.

35

CLAIMS:

I claim:

1   1.   A non-contact scanning laser system for
2   performing corneal refractive surgery by reshaping a
3   portion of a corneal surface comprising:
4   a laser (10, 35) having a pulsed output beam of
5   predetermined ultraviolet wavelength 'and having an
6   energy level less than 10 mJ/pulse;
7   a scanning mechanism (12, 37) for scanning said
8   selected laser output beam (11, 38), said scanning
9   mechanism (12, 37) including a galvanometer scanning
10   mechanism for controlling said laser beam into an
11   overlapping pattern of adjacent pulses;
12   a coupling mechanism (15, 44) coupling said laser
13   beam (11, 38) to a scanning device (12, 37) for
14   scanning said laser beam over a predetermined surface
15   area;
16   focusing optics for scanning said laser beam
17   (11, 38) onto a corneal surface to a predetermined
18   generally fixed spot size;
19   alignment mechanism (17, 43) for aligning the
20   center of the said scanning laser beam onto the
21   patient's eye corneal surface with a visible aiming
22   beam (18);
23   controlling means (23) for controlling the
24   scanning mechanism (12, 37) to deliver the scanning
25   laser beam (11, 38) in a predetermined overlapping
26   pattern onto a plurality of positions on the corneal
27   surface to photoablate or photocoagulate corneal
28   tissue to remove from .05 to .5 microns of corneal
29   tissue per pulse with overlapped pulses to remove

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30   tissue to a desired depth, whereby a low power non-
31   contact scanning laser system improves corneal
32   reshaping surgery.

1   2.   A non-contact scanning laser system in
2   accordance with claim 1 in which the laser (10, 35) is
3   a diode-pumped UV laser having an output wavelength
4   between 193 and 220 nanometers, and energy per pulse
5   of 0.01 to 5 mJ/pulse, a repetition rate of between 1
.6   Hz and 10 KHz, and a pulse duration between 0.1
7   picoseconds to 50 nanoseconds and a focused spot size
8   of (0.05-1.5) mm on the corneal surface.

1   3.   A non-contact scanning laser system in
2   accordance with claim 1 in which the laser (10, 35) is
3   a flash lamp pumped UV laser having an output
4   wavelength between 193 and 220 nanometers, and energy
5   per pulse of 0.1 to 10 mJ/pulse, a repetition rate of
6   between 1 Hz and 10 KHz, and a pulse duration between
7   0.1 picoseconds to 50 nanoseconds and a focused spot
8   size of (0.05-1.5) mm on the corneal surface.

1   4.   A non-contact scanning laser system in
2   accordance with claim 1 in which a laser (10, 35) is
3   an argon fluoride excimer laser having an output
4   wavelength of 193 nanometers, energy per pulse of 0.5
5   to 10 mJ/pulse and a focused generally fixed spot size
6   of between 0.2 to 2 mm on the corneal surface, and a
7   repetition rate of between 1 to 1,000 Hz, and pulse
8   duration of between 1 to 50 nanoseconds.

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1   5.   A non-contact scanning laser system in
2   accordance with claim 1 in which the laser (10, 35) is
3   a free-running Ho:YAG laser having an output
4   wavelength of about 2.1 microns at an average power of
5   between 0.5-5 watts and a focused generally fixed spot
6   size of between 0.1-1 mm.

1   6.   A non-contact scanning laser system in
2   accordance with claim 1 in which the laser (10, 35) is
3   a free-running Er:glass laser having an output
4   wavelength of about 1.54 microns at an average power
5   of between 0.5-5 watts with a focused generally fixed
6   spot size of between 0.1-1 mm.

1   7.   A non-contact scanning laser system in
2   accordance with claim 1 in which the laser (10, 35) is
3   a free-running Er:glass laser having an output
4   wavelength of between 1.9 to 2.5 microns at a power of
5   between 0.5-5 watts and a focused generally fixed
6   spot size of between 0.1-1 mm.

1   8.   A non-contact scanning laser system in
2   accordance with claim 1 in which the laser (10, 35) is
3   a Q-switched Er:YAG laser having an output wavelength
4   of 2.94 microns, and a pulse duration of between 50 to
5   400 nanoseconds, with an energy per pulse of between
6   50-500 mJ and a repetition rate of between 1 and 200
7   Hz with a focused generally fixed spot size of between
8   0.2-2 mm.

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1   9.   A non-contact scanning laser system in
2   accordance with claim 1 in which the laser (10, 35) is
3   an ultra-short pulsed laser having an output
4   wavelength of between 750 to 1100 nanometers, energy
5   per pulse of between 0.01 to 100 microjoules, and a
6   repetition rate of between 0.01 to 100 MHz, and pulse
7   duration of between 0.05-10 picoseconds and a focused
8   generally fixed spot size of between 0.05-0.5 mm.

1   10.   A non-contact scanning laser system in
2   accordance with claim 1 in which the laser (10, 35) is
3   an OPO mid-IR laser having an output of 2.5-3.2
4   microns, a pulse duration of between 1-40 nanoseconds
5   and energy per pulse of between 0.1 to 10 mJ, and a
6   repetition rate of between 10 and 5,000 Hz and a
7   focused generally fixed spot size on the corneal
8   surface of between 0.1 - 2 mm.

1   11.   A non-contact scanning laser system in
2   accordance with claim 1 in which a focusing lens (14,
3   42) for delivering said laser beam (11, 38) is highly
4   transparent to the said laser beam and has a focal
5   length of (50-1500) mm for focusing the laser source
6   onto a generally fixed spot size of 0.05-2 mm on a
7   predetermined position on the corneal surface.

1   12.   A non-contact scanning laser system in
2   accordance with claim 1 in which said controlling
3   means (23) controls said scanning mechanism (12, 37)
4   to scan a pattern of radial aligned spots (Figures 4A,
5   4C, 7A) using a laser beam capable of photocoagulation
6   corneal tissue.

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1   13.   A non-contact scanning laser system in
2   accordance with claim 1 in which said controlling
3   means (23) controls said scanning mechanism (12, 37)
4   to scan a pattern of concentric generally fixed spots
5   (Figures 4A, 4B, 4C, 4D, 6B, 6C) using a laser beam
6   capable of photocoagulating corneal tissue.

1   14.   A non-contact scanning laser system in
2   accordance with claim 1 in which said controlling
3   means (23) controls said scanning to scan a pattern of
4   generally fixed area ring spots (Figures 4A-4E & 6A-
5   6D) using a laser beam capable of photocoagulating
6   corneal tissues.

1   15.   A non-contact scanning laser system in
2   accordance with claim 1 in which said controlling
3   means (23) controls said scanning to scan a pattern of
4   overlapping generally fixed area ring spots (Figures
5   6A-6D) using a laser beam capable of photoablating
6   corneal tissue for myopic correction.

1   16.   A non-contact scanning laser system in
2   accordance with claim 1 in which said controlling
3   means (23) controls said scanning to scan a pattern of
4   overlapping generally fixed area ring spots (Figures
5   6A-6D) using a laser beam capable of photoablating the
6   corneal tissue for hyperopic correction.

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1   17.   A non-contact scanning laser system in
2   accordance with claim 1 in which said controlling
3   means (23) controls said scanning to scan a pattern of
4   overlapping circles of fixed area (Figures 6A-6D)
5   using a laser beam capable of photoablating the
6   corneal tissue for astigmatic correction.

1   18.   A non-contact scanning laser system in
2   accordance with claim 1 in which said controlling
3   means (23) controls said scanning to scan a pattern of
4   radial aligned slits (Figures 7A & 7B) of fixed area
5   using a laser beam capable of photoablating corneal
6   tissue for laser radial keratectomy.

1   19.   A non-contact scanning laser system in
2   accordance with claim 1 in which said controlling
3   means (23) controls said scanning which has a circular
4   scanning pattern to deliver uniform laser energy over
5   a coated window (44) positioning near the corneal
6   surface.

1   20.   A non-contact scanning laser system in
2   accordance with claim 19 in which said scanning
3   mechanism (12, 37) scans a coated window (44) having
4   a predetermined coating to direct said laser beam
5   therethrough and to photoablate the corneal surface to
6   meet a predetermined profile for refractive
7   corrections.

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1   21.   A non-contact scanning laser system in
2 accordance with claim 19 in which said scanning
3 mechanism (12, 37) scans through a coated window (44)
4 made of materials transparent to a UV laser having an
5 output beam of (193-215) nm.

1   22.   A non-contact scanning laser system in
2 accordance with claim 19 in which said scanning
3 mechanism (12, 37) scans through a coated window (44)
4 made of materials highly transparent to an IR laser
5 having an output beam of (2.5-3.2) microns.

1   23.   A non-contact scanning laser system in
2 accordance with claim 1 in which said scanning
3 mechanism (12, 37) scans a uniform scanned pattern
4 (Figures 9A & 9B) with a spatial overlap of 50-80%
5 and beam orientation whereby the initial beam profile
6 uniformity is not critical.
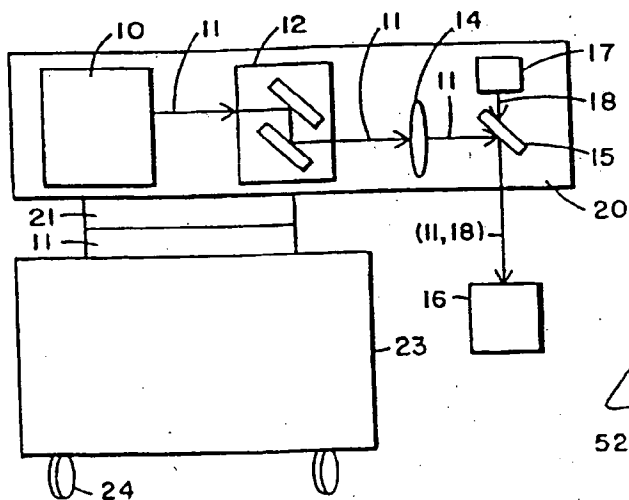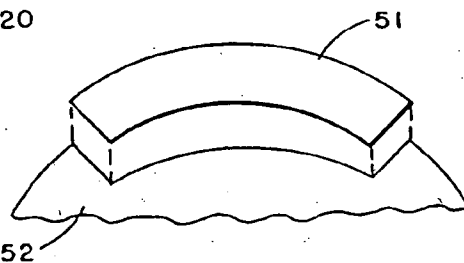
FIG. 1

FIG. 5A
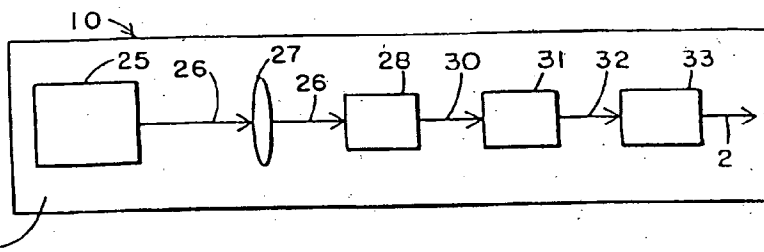
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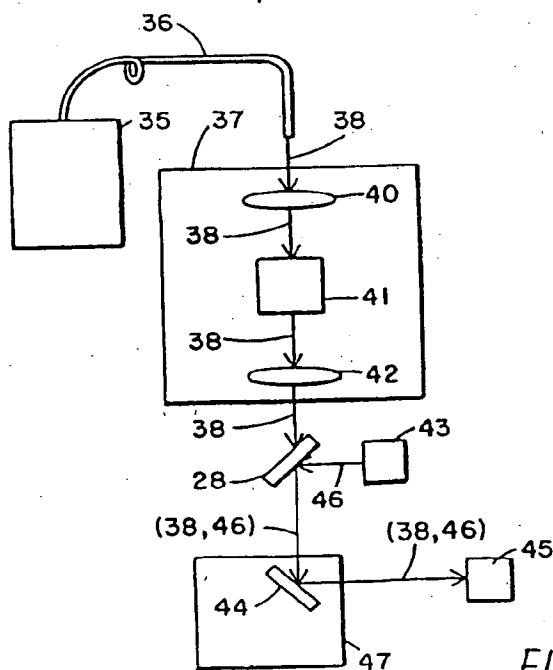
FIG. 3
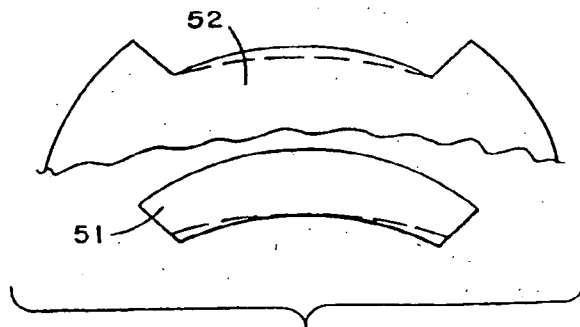
FIG. 5B
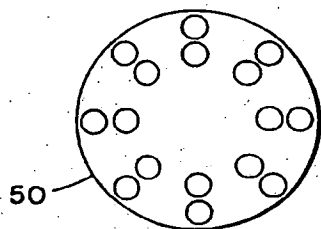
FIG. 4A
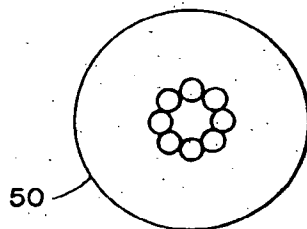
FIG. 4B
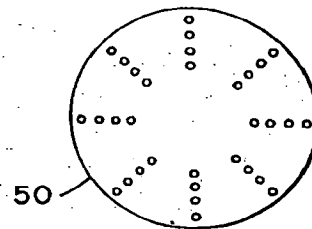
FIG. 4C
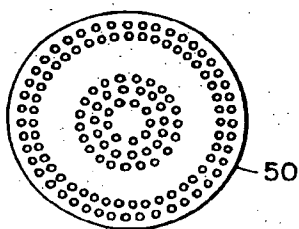
FIG. 4D
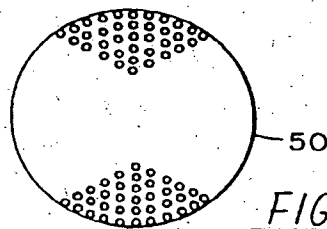
FIG. 4E
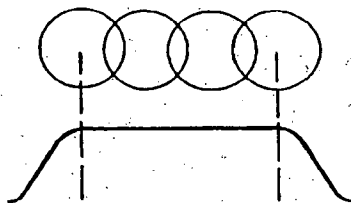
FIG. 6A
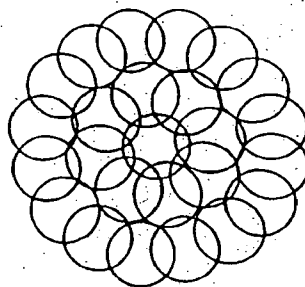
FIG. 6B
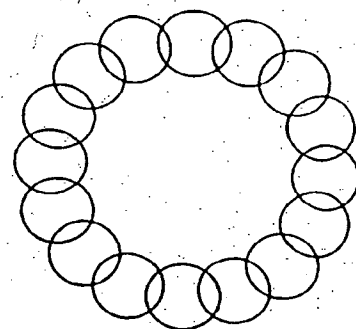
FIG. 6C

FIG. 6D

FIG. 7A   _FIG. 7B_

FIG.   8A
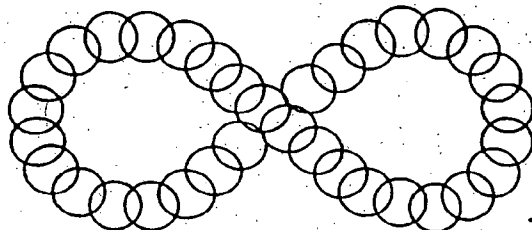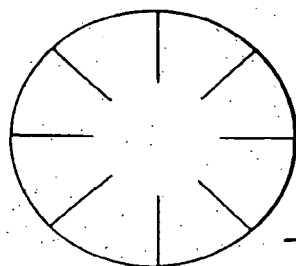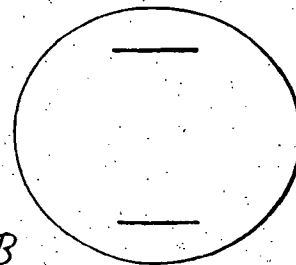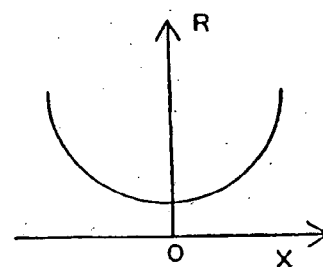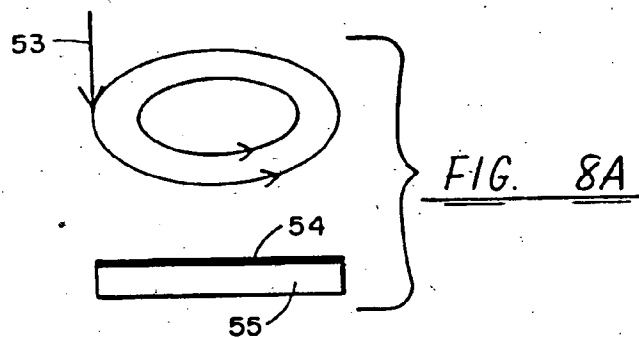
FIG.   8B
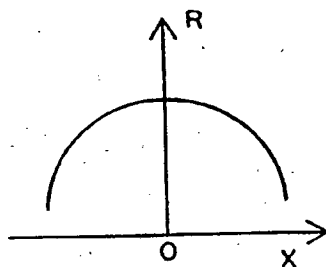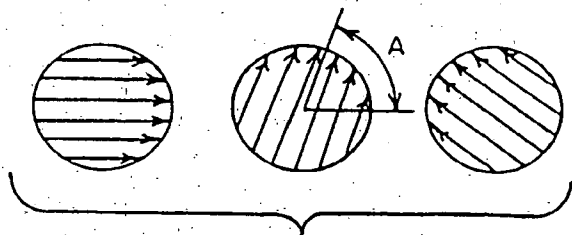

FIG.   8C
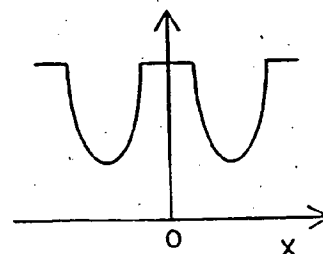
FIG.   8D
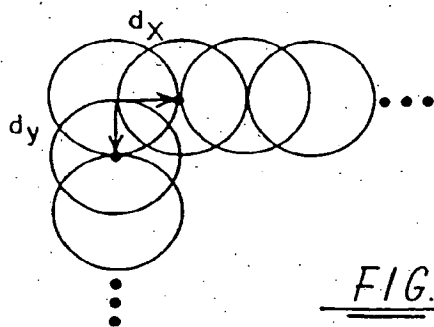
FIG.   9A
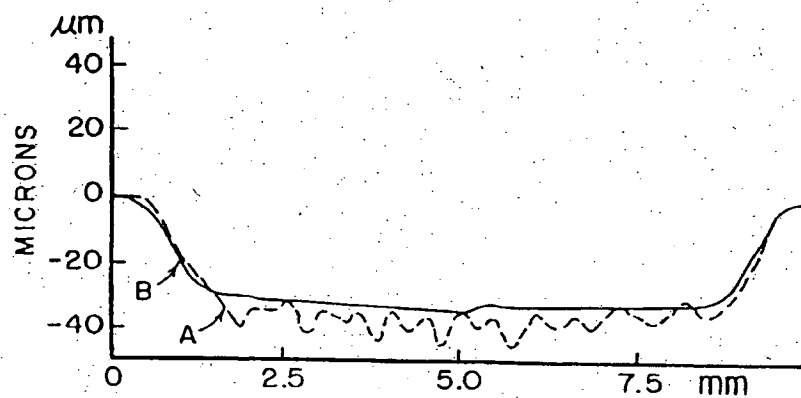
FIG.   9B

FIG.   10A
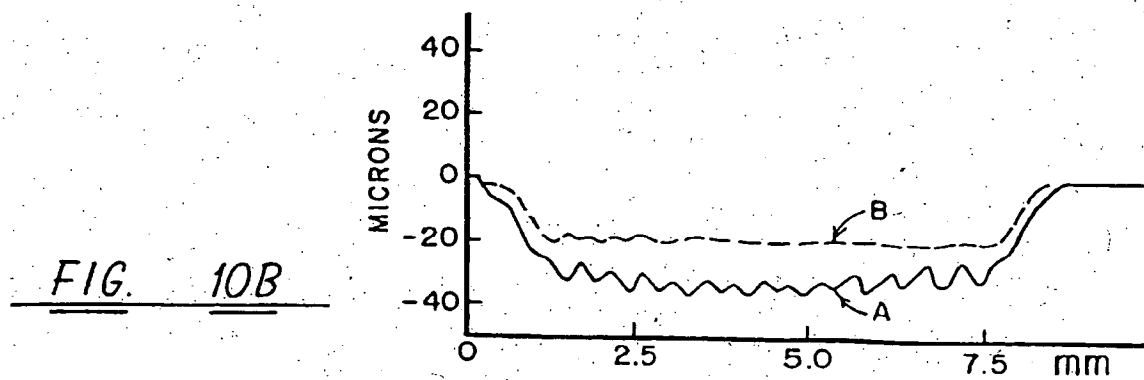
FIG.   10B
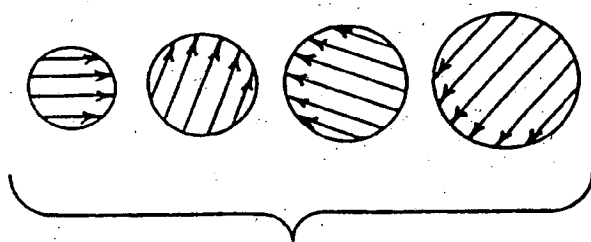
FIG. 11A
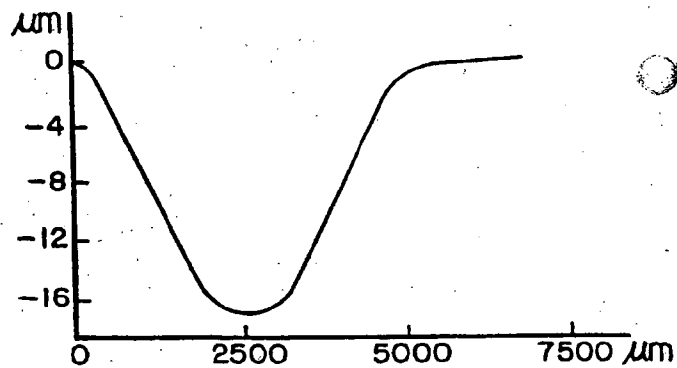
FIG. 11B

INTERNATIONAL SEARCH REPORT

A. CLASSIFICATION OF SUBJECT MATTER

IPC(6) :A61N 5/02

US CL :606/005

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

U.S. : 606/3-6, 10-12

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

NONE

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

NONE

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	US, A, 4,718,418 (L'ESPERANCE, JR.) 12 January 1988, see whole document.	1-23
A	US, A, 4,729,372 (L'ESPERANCE, JR.) 08 March 1988, see whole document.	1-23

☐ Further documents are listed in the continuation of Box C.   ☐ See patent family annex.

*	Special categories of cited documents:	"T"	later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
"A"	document defining the general state of the art which is not considered to be part of particular relevance		
"E"	earlier document published on or after the international filing date	"X"	document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
"L"	document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)		
		"Y"	document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art
"O"	document referring to an oral disclosure, use, exhibition or other means		
"P"	document published prior to the international filing date but later than the priority date claimed	"&"	document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
30 MAY 1996	08 AUG 1996

Name and mailing address of the ISA/US	Authorized officer
Commissioner of Patents and Trademarks Box PCT Washington, D.C. 20231	MICHAEL PEFFLEY
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